MAGNETIC RESONANCE IMAGING (MRI) AND SPECTROSCOPY (MRS) USING SIMULTANEOUS 2-CHANNEL ACQUISITIONS: APPLICATION FOR MOUSE BRAIN EXAMINATION BY RECONFIGURATION OF A "STANDARD" BRUKER SPECTROMETER

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ABSTRACT

In the field of small animal imaging, the interest for phased array coil imaging is growing but high field MR experimental systems with multiple receiver channels are still rare and the upgrade of existing systems is relatively expensive. In this work, a standard 4.7 T Bruker Biospec Avance II spectrometer was modified to allow simultaneous two-channel acquisitions. Modifications were validated on imaging and spectroscopy on metabolite solution phantom as well as on mice brain using a home-made two-channel array coil operating at 200.3 MHz. A dedicated two-channel array coil with two square elements encompassing the mouse brain was designed and built. Compared to a single-channel surface coil, the mean SNR measured on images in the ROI corresponding to whole mouse brain was improved by about 30% as well as the signal uniformity. For spectroscopic acquisition, the SNR gain in a voxel located close to the coils was improved by about 65%. Modifications realized for proton multiple-channel acquisitions could also be applied for any X-nucleus. Compared to quadrature detection coils, two-channel coils offer the ability to use parallel acquisitions techniques.

Index Terms— Magnetic Resonance Imaging, Magnetic Resonance Spectroscopy, Signal to Noise Ratio, Array Coil, Small Animal

1. INTRODUCTION

The advantages of array coil imaging on clinical MR systems have gained relevance in many applications. In the field of small animal imaging, the interest for this concept is growing. Array-coil imaging is an advanced method to enhance the signal-to-noise ratio (SNR) using superior sensitivity of several small coil elements compared for example to a larger single element covering the same field of view (FOV) [1, 2]. Two or more elements of small diameter are placed side by side on the same coil. Every single element has its own channel for signal reception. Thus, multiple images (and also k-spaces) are obtained from every element and then combined in one "array" image. This technique requires high computer speed to reconstruct the individual and combined images [2-4]. Another benefit of phased-array technique is the possibility to use parallel imaging techniques such as SENSE (sensitivity encoding) and GRAPPA (generalized autocalibrating partially parallel acquisitions) to increase acquisition speed [5, 6].

High field MR experimental systems with multiple receiver channels are still rare and the upgrade of existing systems is relatively expensive. In this work, a "standard" 4.7 T Bruker Biospec Avance II spectrometer with two broadband chains, one dedicated to proton (^{1}H), one dedicated to X nuclei (with X=^{31}P, ^{23}NA, ^{13}C, ...) was modified to allow two-channel ^{1}H acquisitions.

These modifications were validated *in vitro* with phantoms as well as *in vivo* on mice brain for imaging and spectroscopy using a home-made two-channel phased array coil operating at 200.3 MHz.

2. MATERIAL AND METHODS

2.1. Array coil

The experiments were performed on a Bruker 4.7T Biospec system (Bruker, Ettlingen, Germany), the maximum gradient amplitude available is 270 mT/m and the clear bore diameter is 100 mm. A linear 72 mm inner diameter birdcage coil (Rapid Biomedical, Würzburg, Germany), for excitation, and a home designed two-channel phased array coil, for reception, were used.

The two-channel array coil was built on a Plexiglas™ cylinder with 21 mm outer diameter and 19 mm inner diameter. Each element consists in a rectangular single loop with 12 x 16 mm^2 internal and 15 x 20 mm^2 external dimensions (Figure 1). One leg of this loop is common for the two elements. The S-parameters (S_{11}, S_{22}, and S_{12}) and the quality factor (Q) were measured with an Agilent ENA300 Network Analyzer (Agilent Technologies Inc., Santa Clara, CA, USA). The S-parameters describe the performance of a multiple-port network by measuring the traveling waves that are scattered or reflected when a network is inserted into a transmission line of a certain characteristic (in our case 50 Ohms). The quality factor is a

measure of the ratio of the energy stored in the resonant circuit to the energy lost during one period at working frequency.

The decoupling of the two channels (the compensation of the mutual inductance between both channels) was achieved using a fixed decoupling capacitor on the common conductor of the two loops. The value was experimentally adjusted to minimize the S_{12}, parameter between the two channels. Both channels were tuned at 200.3 MHz corresponding to the proton's resonance frequency at 4.7 T and matched to 50 Ω for this frequency using non-magnetic case A series 100 and 710 ATC capacitors (American Technical Ceramics, New York, USA).

The fine tuning/matching and actively decoupling circuit was designed to be interfaced and driven by the "Bruker Decoupling Box". For fine tuning and matching, variable capacitance diodes BB 149 (Philips Semiconductors, Eindhoven, Netherlands) driven with variable voltage were used. For decoupling the array coil from the emitting coil, both channels of the coil had an active decoupling circuit with two parallel DH 80106 PIN diodes (Temex Ceramics, Pessac, France) (Figure 1).

The home-designed phased-array coil was compared to a single channel 15 mm diameter multipurpose surface coil.

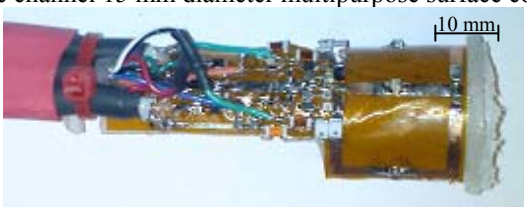


Figure 1. Photograph of the home-designed two-element phased array receive coil. Each inner coil element is 12 x 16 mm^2. This coil operates at 200.3 MHz.

2.2. Bruker spectrometer

For a standard proton acquisition the routing which defines the connections between the hardware parts involved in the acquisition pipeline is show in Figure 2.

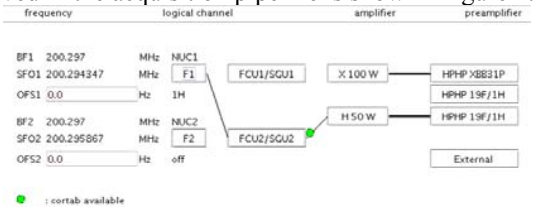


Figure 2. The routing connection scheme used for a standard (one channel) proton acquisition.

To perform a simultaneous two-channel acquisition, the routing of the MRI spectrometer was modified according to Figure 3. The excitation was performed using the X nucleus amplifier, which is a broadband amplifier. The output of the amplifier was linearized, for the proton frequency, by creating a correction table (cortab). This cortab file is used to correct the non-linearity of the RF pulse power level versus the pulse length.

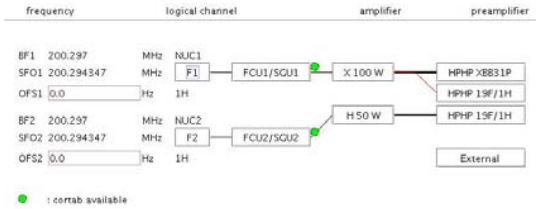


Figure 3. The routing connection scheme used for a simultaneous two-channel proton acquisition.

For reception the standard ^{1}H-nucleus chain together with the X-nucleus chain, which was interconnected with a second proton preamplifier, were used. The methods including the pulse programs were modified to activate the second receiver channel and to handle the sum of squares reconstruction (only for imaging). For imaging the MSME, FLASH and RARE sequences and for spectroscopy the PRESS sequence, were modified for simultaneous two-channel acquisition.

2.3. *In vitro* protocol

For *in vitro* imaging and spectroscopy experiments, a 18 mm in diameter cylindrical phantom made of eleven metabolite (aspartate (Asp), creatine (Cr), choline (Cho), γ-aminobutyric acid (GABA), glucose (Glc), glutamate (Glu), glutamine (Gln), N-acetylaspartate (NAA), taurine (Tau), lactate (Lac) and myo-inositol (Ins), 50 mM, pH = 7, 10 ml) in aqueous solution was used [7]. The signal uniformity, the possible artefacts from electronic components, the efficiency of the decoupling methods and the SNR were then compared with the reference 15 mm diameter receive-only surface coil.

A T1-weighted Spin-Echo (SE) sequence was used with the following parameters: FOV = 35 x 35 mm^2, 256 x 256 matrix size, 16 slices of 2 mm thickness, 50 kHz receiver bandwidth (rbw), 260 ms repetition time (TR) and 10.7 ms echo time (TE). The SNR (mean ± standard deviation) was calculated on the Region of Interest (ROI) defined as an equivalent area corresponding to the surface and the location of the mouse brain (Figure 5B).

Spectroscopic acquisitions were performed using a short–echo time PRESS sequence (TE/TR = 20/5000 ms, Tacq = 21 min, 4096 data-points, bandwidth of 4 kHz). The volume of interest (2.5 x 2 x2 mm^3) was located as shown in Figure 5A. First- and second-order shim terms were adjusted using FASTMAP. The spectroscopy data from the two channels were combined using Matlab 7.4 (Mathworks Inc, Natick, MA, USA) in the time domain using a sum of squares weighting function. Prior to the combination, the signals from the two channels were zero-order phase corrected. The coil intensity weighting factor, for each coil was obtained from the mean value of the four first absolute time-domain data points of the unsuppressed-water signal. The SNR was measured for three specific metabolites (Cho, Cr, NAA).

2.4. *In vivo* protocol

In vivo experiments were performed on mouse brain. The ethical guidelines for experimental investigations with animals were followed. Gaseous anaesthesia was performed on mice placed in prone position. The home-designed phased-array coil and the 15 mm surface coil were placed on top of the brain. An axial T2-weighted fat suppressed (FS) RARE sequence was used with the following parameters: TR/TE = 4000/75 ms; RARE factor = 8; 30 x 30 mm^2 FOV, 256 x 192 matrix, 19 slices, 1 mm slice thickness and 17 kHz receiver bandwidth.

3. RESULTS

The measured quality factor of the unloaded coil was about 139 for every single channel. For a loaded coil with a 17.5 mm diameter cylinder phantom (filled with 0.45 % NaCl solution), the quality factor decreased up to 127. The decoupling capacitor mounted on the circuit was about 56 pF. The isolation (decoupling) between the two channels measured with the S_{12} parameter was 29 dB (Figure 4).

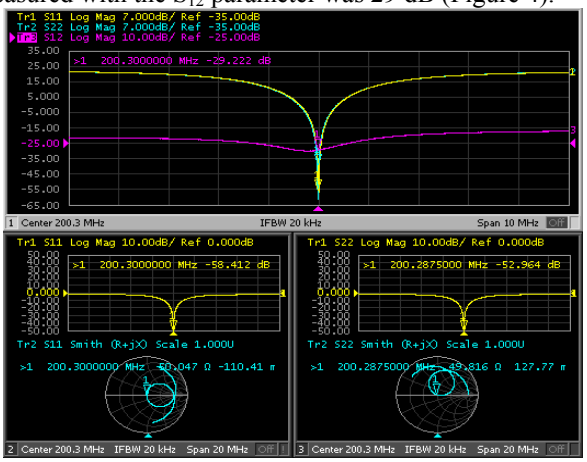
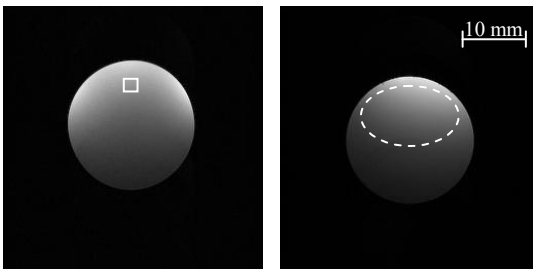


Figure 4. The S-parameters (S_{11}, S_{22}, and S_{12}) measured with a network analyzer. The upper zone shows the two channels, S_{11} and S_{22}, and the transfer between channels S_{12} which show attenuation higher then 29 dB. The down zone shows the magnitude and Smith diagram for both channels (left for S_{11} and right for S_{22}).

No visible susceptibility artefacts from the components were depicted on images obtained with the uniform cylindrical phantom (Figure 5). The SNR gain for the two-channel array coil was up to 1.3 compared to the SNR obtained with the 15 mm diameter surface coil. In addition, the intensity signal measured in the ROI (Figure 5B) was more uniform with a SNR standard deviation of 6 compared to 21 measured on images acquired with the surface coil.

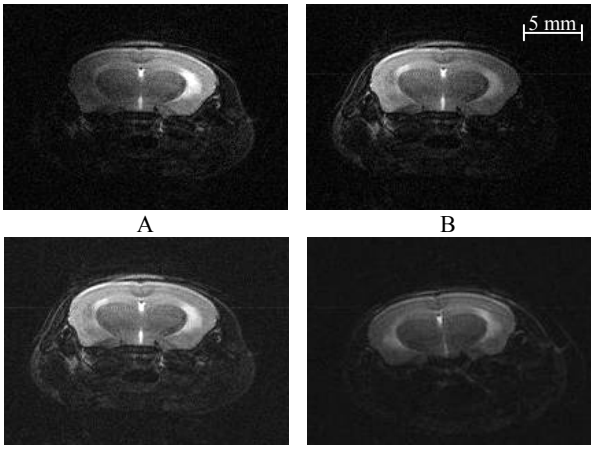
The images acquired *in vivo* on a mouse brain, with neurological disorder, using the two-channel phased-array-coil (before and after combination) and the surface coil are shown in Figure 6.



A B

Figure 5. Axial MR images obtained on a homogenous cylindrical phantom with (A) the two-channel phased-array coil and (B) with the 15 mm surface coil. The location of the PRESS volume is shown in (A). The ROI used for imaging SNR comparisons is shown in (B).

The SNR was measured in various locations on the mouse brain images (hippocampus, cortex, LCR) and the mean value was 137 ± 15 for the two channel array-coil. For the 15 mm single channel surface coil the mean SNR was 117 ± 41.



A B C D

Figure 6. (A,B,C) Axial T2-weighted images acquired on a mouse brain with neurological disorders using the home-designed phased array coil (A,B) before and (C) after the combination of the two channels. (D) Axial T2-weighted images acquired on the same mouse with the 15 mm diameter multipurpose surface coil.

The spectra acquired by the two-channels before and after combination are shown respectively in Figure 7 and in Figure 8.

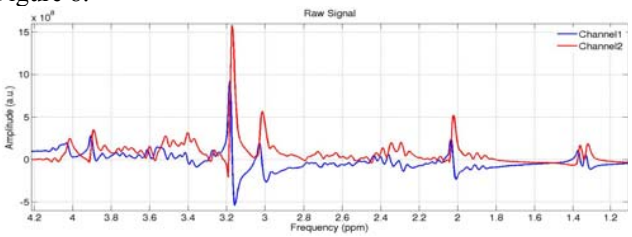


Figure 7. Acquisitions performed with the home-designed phased array coil; Spectrum for each channel acquired on the eleven metabolite solution phantom.

The SNR measured for three metabolites of interest, after the combination of the two-channel compared to the surface coil are summarized in Table 1.

Metabolites	Phased-array	Surface
Cho	317	192
Cr	111	69.2
NAA	113	65.9

Table 1. Comparison of SNR-values measured *in vitro* for three metabolites using a two-element phased-array and a surface coil respectively.

Magnetic field homogeneities measured with both coils were comparable with a full width half maximum (FWHM) of 3.2 Hz (ranging from 2 to 5 Hz) indicating that the home-designed coil do not affect the spectra resolution.

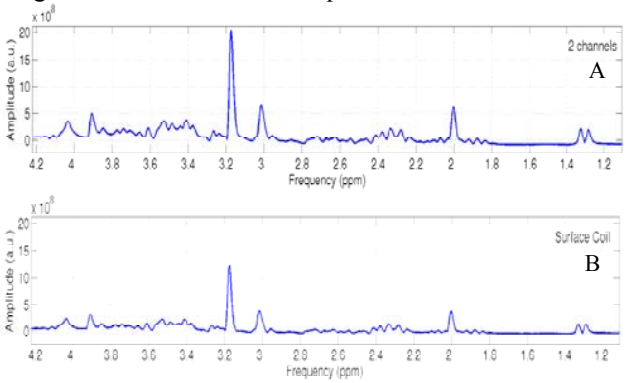


Figure 8. Spectra acquired on the eleven metabolite solution (A) after combination with the two-channel coil; (B) with the surface coil.

4. CONCLUSION

The isolation measured between the two channels of the phased array coil validates that the shared inductor method use to eliminate the coupling between elements was efficient. The capacitance inserted in the common conductor of the two resonant circuits can be easily adjusted to offer the maximum level of isolation between the two channels. This shared inductor decoupling method is particularly suitable to decouple small size elements yielding to both good sensitivity and parallel imaging capabilities [8-10]. This decoupling method is however not extendable to elements with no shared conductors. This is usually the case for coils with more than two channels for which a geometrical overlap design is more appropriate [9, 11, 12]. Interfaced with the Bruker Decoupling Box, the phased array coil was easy to adjust (tuning and matching) with different loading conditions (samples or mouse head).

The SNR-value measured on combined images increase by a factor of 1.3. The curve geometry used for the phased-array contributed to the signal uniformity improvement in the ROI.

Simultaneous two-channel acquisition for imaging and spectroscopy was demonstrated after reconfiguration of a standard 4.7T Biospec spectrometer. Only a second broadband RF amplifier in the proton range is mandatory. The second proton preamplifier can be optional if the two-channel amplifier decoupled phased-array coil with on board preamplifiers is used. Modifications realized for proton multiple channel acquisitions could also be applied for any X-nucleus. Compared to quadrature detection coils, two-channel coils offer the ability to use parallel acquisition techniques. Addition of array technique with quadrature combination can be considered in order to build a circularly polarized two channel array coil to maximize the SNR.

The interest of the home-designed coil will be validated for *in vivo* experiments prior to be applied for spectroscopic monitoring of neuronal diseases progression on mice models. Further step will be to implement parallel acquisition techniques and to demonstrate the interest for small animal ^{1}H imaging as well as for other nuclei such as ^{3}He hyperpolarized gas imaging.

5. ACKNOWLEDGEMENTS

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6. REFERENCES

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